

U.S. DEPARTMENT OF ENERGY
FEDERAL ASSISTANCE PROGRAM/PROJECT STATUS REPORT

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4. Name and Address David M. Hamby Dept of Nuclear Engineering and Radiation Health Physics Oregon State University Corvallis, OR 97331 -5902		5. Program/Project Start Date June 1, 2002
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7. Approach Changes The prototype triple-layer phoswich detector has been replaced with a newly innovative design beta spectrometer/dosimeter that couples scintillating plastics with a photodiode. This arrangement has not yet been attempted for beta dosimetry and beta spectroscopy. <input type="checkbox"/> None		
8. Performance Variances, Accomplishments, or Problems The new prototype is functional and PhD student, Aaron Kriss, is in the process of characterizing its performance. <input type="checkbox"/> None		
9. Open Items Our 2 nd PhD student, Lan Lin, is having trouble with her entrance visa and with the SARS outbreak. Therefore, a new graduate student will be hired this summer to handle deconvolution and neural network research. <input type="checkbox"/> None		
10. Status Assessment and Forecast Even though we've switched detector designs, and had setbacks with Chinese students, we're well on track for completion of the project as originally proposed. <input type="checkbox"/> No Deviation from Plan is Expected		
11. Description of Attachments Attached is the Preliminary Exam proposal submitted, defended, and passed by Aaron Kriss. <input type="checkbox"/> None		
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Table of Contents

Chapter 1: Overview	3
Introduction	3
Objectives	6
Significance of the research	6
Research Questions	6
Definitions of Terms	7
 Chapter 2: Review of the Literature	 8
Avalanche photodiodes (APD) and large area avalanche photodiodes (LAAPD) for use in radiation detection and spectroscopy	8
Theoretical beta dosimetry	13
Beta dosimetry and spectroscopy with plastic scintillators	14
 Chapter 3: Materials and methods	 18
<i>Materials</i>	18
Large area avalanche photodiode and electronics	18
Scintillators	20
Housing	22
Sources	22
<i>Methods</i>	24
Characterization of the LAAPD module	24
Characterization of the scintillator	25
Calibration of the system	25
Cerenkov component	25
Theoretical dosimetry	26
Theoretical spectroscopy	26
Experimental dosimetry	26

Experimental spectroscopy	27
Conclusion	27
References	28
Appendix: Schedule	34

Development of a beta dosimeter and spectrometer utilizing plastic scintillator volumes and a large area avalanche photodiode.

Chapter 1: Overview

Introduction

After Becquerel observed the mysterious fogging of his photographic plates in 1896 when they were in close proximity to, but opaquely separated from, uranium salts, he postulated the existence of some new “radiation effect”, unrelated to any other known sort: i.e., radioactivity (Brown 2002, p 12). Becquerel had made a momentous discovery, but he didn’t really know what was occurring. How could he have? Roentgen had only discovered x rays the year before; J. J. Thomson at the Cavendish Laboratory would not definitively discover the electron until the following year (Brown 2002, p 13). It wasn’t until Rutherford and Soddy made the revolutionary assertion early in the 20th century that Becquerel’s mysterious radiation was in fact the product of the transformation of certain elements into different elements that an understanding of the phenomenon of radioactivity was born (Brown 2002, p 47). The doors to discovery were wide open. So were the doors to a whole new kind of health hazard.

Roentgen’s x ray discovery took very little time to make its way around the globe, and very little more to become a medical tool. In the meantime, basic research related to radioactivity took center stage in the work of the Curies and others. The research took place in an environment where little or no thought was given to safety, not because the

researchers were careless, but because they weren't aware of ionizing radiation's biological hazards. There were clues, though.

Researchers had noted the ulcerating effect of x-rays after 1895 (Brown 2002, p 152). In 1903 Rutherford noticed that the hands of Pierre Curie, as he held a flask of glowing radium, were scarred. Curie had a difficult time handling the flask due to the radiation burns. Becquerel also suffered a chest burn from radium in a small flask, given to him by the Curies, he had held in a vest pocket for a length of time. Radium is an intense gamma emitter, and it is the gamma phenomenon that initially drew all the attention related to safety. In many ways, at least for external exposures, this is still true. It's not hard to understand why: the deep penetration capabilities of gammas and x-rays make them the most important of the non-neutron radiations in matters of external safety. Under certain conditions, however, short-range charged particle radiations, such as the beta (a name given the then-mysterious charged radiation by Rutherford before 1900), become important in external radiation protection.

Protection against betas is usually a simple matter of interposing a thin layer of low-Z material between source and target. But if we want to measure beta dose rates, for example from an accident, or identify unknown radionuclides, we need instruments specially built for the purpose.

This amounts to quantifying, in some way, the energy of the beta and where it deposits its energy. There is any number of ways to do this (Knoll 1989). One approach is to analyze

the light produced when beta particles traverse a scintillator. With the proper tools and analysis, the scintillator light can be employed to determine not only the beta spectrum, and thus the radionuclide's identity, but also the dose delivered by the betas to some mass of interest.

Dose from ionizing radiation is defined as energy imparted to mass, divided by that mass (Attix 1986). By building a scintillator with specific dimensions corresponding to a mass distribution of interest, for example a layer of skin, dose to that mass can be inferred by quantifying the light output from the scintillator. From the light output the energy imparted to the scintillator is known, and if the scintillator has characteristics that are closely tissue-equivalent (density close to water and atomic composition near to that of tissue), then dose to the mass of interest is simply energy divided by mass. Since plastic scintillators meet the requirements of tissue equivalency and are easily shaped to any specification, they will be used in the dose analysis part of this research.

Plastic scintillators are also useful for beta spectroscopy and will be employed here for that purpose. However, since the scintillator light will be detected by a solid state device suitable by itself for beta detection, namely a large area avalanche photodiode, we will also pursue experiments using the photodiode directly for beta spectroscopy.

By utilizing specially built scintillators and tools, this research will extend the current level of capability for analyzing beta spectra and beta dose.

Objectives

Our goal is to enhance current beta particle dosimetry and spectroscopy methods by development of a new, dual-purpose beta detector. The device will serve alternately as a dosimeter and a spectrometer, by utilizing near tissue-equivalent plastic scintillator material in precise configurations, coupled to a large area avalanche photodiode module. The photodiode will also potentially serve as a direct beta interaction spectrometer, depending on the outcome of the current research.

Significance of the research

Beta particles take a back seat to gamma rays in the area of external radiation protection, for the simple reason that betas are far more easily protected against. Thus, betas pose a lesser hazard. However, there are cases in which betas constitute a danger. For example, large-scale accidents on the order of Chernobyl may result in considerable beta emitter distribution, as may a terrorist “dirty” bomb. In these cases, the ability to quickly acquire an accurate dose estimate, as well as an analysis of the fallout’s composition, will be of significant value. Our device is designed to provide this capability.

Research Questions

Can we develop a beta dosimeter using a volume of scintillator, coupled to a special photodiode module, which will accurately predict dose to tissue?

Can we develop a beta spectrometer, using a large area avalanche photodiode, which will accurately and quickly allow us to identify a mixture of beta emitting radionuclides?

Definitions of Terms

Large area avalanche photodiode (LAAPD): A semiconductor device reverse biased at high voltage so as to cause a multiplication process when an electron-hole pair is created by some incident photon or other radiation event. Acts like a photomultiplier tube for radiation detection, but is much smaller and more rugged.

Scintillator: A substance, either organic or inorganic, liquid, solid, or gas that emits light in a spectrum defined by the character of the substance, when impinged upon by some sort of radiation. The amount of light is related to the energy imparted to the substance and can be analyzed in a variety of ways, depending on application.

Beta particle: A form of ionizing radiation, physically an electron that is emitted by some radioactive elements. The energy of each beta is governed by statistics peculiar to each element.

Waveshifter: A compound that absorbs light of one wavelength and re-emits it at another wavelength.

Conversion electron: A shell electron ejected by an atom as an alternative to gamma emission.

Cerenkov effect: Light produced as relativistic electrons, traveling faster than the speed of light in whatever the electron is traveling through, slow down. Significant in water, plastic and other transparent materials.

Characteristic x-ray: An x-ray emitted from an atom when an orbital electron drops from a higher energy level to a lower energy level.

Energy resolution: The ability of a detector to resolve energy peaks in a spectrum. Usually specified as a percentage of the peak energy by the FWHM of the peak. Low percentages, therefore, mean better resolution.

Gain: A number indicating how much a signal's amplitude is multiplied. A gain of 100 might mean that one original electron becomes 100 electrons at the end of an avalanche process, for example.

Noise: Any distortion of the signal being measured that tends to limit an instruments utility.

Chapter 2: Review of the Literature

Avalanche photodiodes (APD) and large area avalanche photodiodes (LAAPD) for use in radiation detection and spectroscopy

McKay and McAfee, of Bell Labs, gave the first account of an electron avalanche process occurring in a semiconductor solid in 1953 (1953). They observed that by creating a high electric field across a p-n junction electron multiplication took place after injection of an electron. Similar multiplication took place of electrons produced by the photoelectric effect. Though not explicitly interested in possible radiation detection applications of their new device, part of the experimental procedure involved bombarding the semiconductor with alpha particles, with the objective of determining the timing characteristics of the avalanche process. Thus, even in the very earliest days of the solid-state era, avalanche photodiodes had played a role, however inadvertent, in radiation detection.

Johnson, of Texas Instruments, provided evidence that the signal-to-noise and noise-equivalent-power characteristics of diodes biased to breakdown levels appeared to provide the “solid-state analog of the photomultiplier tube” (1965). Again, the focus of his research was not on radiation detection, but rather on communications technology. However, given the importance of the photomultiplier tube in radiation detection, any device that behaved similarly had potential in that field.

McIntyre, of RCA Victor Company, further analyzed the noise characteristics of avalanche diodes (1966). His effort was directed toward understanding the noise

generated during the multiplication process, from which he derived an expression for the noise spectral density, for any distribution of injected carriers.

Locker and Huth, of General Electric Company's Space Sciences Laboratory, first utilized avalanche diodes for radiation detection in 1966 (1966). Their work involved detecting low-energy x rays and protons directly in the diodes, which were only 0.1 cm squared in area. By using suitable electronics they were able to detect x-ray energies as low as 1.49 keV. The authors, with McKinney, extended their research in 1968 to developing a germanium avalanche diode specifically for detecting x rays in the energy range 10-30 keV (Huth et al. 1968). Their main observation was on the difficulty of achieving uniform multiplication across the face of the diode, a problem of obvious importance if the diode areas were to be increased significantly.

Webb and McIntyre studied large area reach-through avalanche photodiodes as applied to x-ray spectroscopy at room temperature, limiting themselves to energies between 1.5 and 20 keV (1976). The diode sizes studied were 12 mm² and 25 mm² areas, small by current standards, but large for the time. Gain was limited to about 63, and energy resolution was at best 600 eV FWHM for ⁵⁵Fe. The various contributors to the noise were studied, such as dark current and gain non-uniformity.

Gelezunas et al. studied avalanche photodiodes with areas of 20 and 330 mm² (1977). Their focus was on an improved fabrication process for producing uniform gain across the area of the diode, a crucial factor governing the usefulness of large-area devices. They

achieved a gain variation of about $\pm 7\%$ across the diode faces. In addition, their device was of the p-n junction type, as opposed to the reach-through type of Webb and McIntyre, so much larger gains of 400 to 800 were attainable, though at a sacrifice of energy resolution.

Reiff et al. covered the basic advantages of large area avalanche photodiodes over photomultiplier tubes (1983). These include ruggedness, high quantum efficiency, light weight and small size, and insensitivity to external magnetic and electric fields. The authors describe their experiments with a 1-inch squared area APD, which they used both coupled to a BGO crystal and directly exposed to low-energy x-rays. It was speculated that such APDs might replace the PMTs in PET scanners, a circumstance that eventually came to pass.

By 1989, Lecomte et al. had developed BGO-APD arrays for PET scanners (1989). They achieved energy resolutions of between 16 and 20% for 0.662 MeV gamma rays. In addition, they demonstrated that when coupled to fast plastic detectors, APDs could be used successfully in fast timing measurements.

By the 1990's, research with and about APDs accelerated. Baron and Ruby (1994) investigated the time responses of x-rays detected directly in large-area APDs of various sizes. Of particular value is their description of how the response varies depending on which part of the APD the x-ray deposits its energy. Response times were typically several nanoseconds. Lorenz et al. made similar studies (1994). Schmelz et al. extended

the use of APDs for PET scanners, achieving spatial resolutions of 2.3 ± 0.1 mm, energy resolution of 15%, and time resolution of 2.6 nsec (1995).

From the mid-90s until the present, a significant number of papers directly concerned with the LAAPD design being used here at OSU began to be published. Ochi et al. wrote favorably regarding their experiments with a 16 mm diameter LAAPD from Advanced Photonix (1996). They studied performance characteristics such as gain dependence on temperature, and timing. X-ray interactions were examined in CsI(Tl) scintillators coupled to the APDs, as well as directly in the LAAPD. Results were good, with energy resolution as low as 4% for small scintillators. Important results were also obtained proving the strong dependence of gain on ambient temperature; consequently, for stable operation, the LAAPD's temperature must be monitored and held constant, if possible.

Moszynski and various collaborators have been particularly productive in studying LAAPDs from Advanced Photonix. Timing studies involving LSO (lutetium oxyorthosilicate) crystals coupled to LAAPDs of 10 and 16 mm diameter were carried out (Moszynski et al. 1996). Response times of less than 20 ns were measured at a gain of 300. In addition, quantum efficiencies of 55% for the 16 mm diode and 65% for the 10 mm diode were observed. LAAPDs can be configured either with a window or without; the researchers found that the number of e-h pairs produced in the APD was reduced by a factor of 2 with the window in place. For that reason we have chosen to use a windowless version for our work.

Since many scintillators emit light in the UV to blue range, and most APDs are preferentially sensitive to longer wavelengths, development of a blue-sensitive APD would be useful. Just such a blue-enhanced LAAPD was examined (Moszynski et al. 1997). The characteristics of the device were found to be just as good as the non-enhanced versions. Though our scintillator of choice is a long wavelength variety, the blue-enhanced LAAPD is still a better match than the non-enhanced type. Therefore, we chose to use a blue-enhanced version in our research.

Further work with LAAPDs from Advanced Photonix has continued, including very recent work (Moszynski et al. 1998), (Moszynski et al. 2000), (Moszynski et al. 2002), (Moszynski et al. 2002), (Belogurov et al. 2003). The emphasis has gradually shifted from examining the LAAPDs themselves, to simply using them for work on such things as characterizing exotic scintillators at low temperatures. The success in these instances gives us confidence that we will be able to utilize an LAAPD in our work.

Lest we be swayed unduly by the work of one group, we can note the use of APDs and LAAPDs by other individuals and groups. Pansart examined using an APD in high-energy physics experiments (1997). Solovov et al. used an LAAPD to examine detection of scintillation light from liquid xenon (2000). Allier et al. did the same for a $\text{LaCl}_3(\text{Ce}^{3+})$ scintillation crystal (2002). Renker reviewed the properties of APDs and LAAPDs for uses such as high-energy physics, astrophysics and medical imaging (2002). Shi et al. studied APDs for use in a tokamak, as soft x-ray detectors (2002). Finally, Rafecas et al.

incorporated APDs into their small animal PET scanner, the aptly named MADPET-II (2003).

For a general consideration of semiconductor radiation detectors, see Knoll (1989). For in-depth coverage of semiconductor devices of all sorts, see Sze (1981).

Theoretical beta dosimetry

The ICRU categorizes theoretical beta dosimetry by technique. These include analytical or deterministic methods, Monte Carlo methods, tables of absorbed dose distributions, derived computer methods such as VARSKIN, and empirical expressions for absorbed dose distributions (ICRU 1994). Loevinger developed empirical formulas for beta dose distributions around point sources based on apparent attenuation coefficients and the average and maximum beta energies (1950, 1954, 1956). Cross extended this work by more accurately modeling the parameters that defined Loevinger's method (1997).

Berger derived point kernels for determining beta dose for situations such as immersion in a radioactive cloud (1974). A number of authors studied beta dose, or more properly electron dose, using Monte Carlo methods (Gualdrini et al. 1994, Hirayama 1994). Fell put forth a method for calculating skin dose based on the continuous slowing down approximation (CSDA), from which he derived a semi-empirical point dose function (1991). Kocher and Eckerman formulated electron dose-rate conversion factors for skin exposure for a variety of geometries, based on Berger's work (1981).

Beta dosimetry and spectroscopy with plastic scintillators

Theoretical dose calculations are useful for purely predictive or retrospective analyses, but for real-world situations there must be measurement as well. We can roughly divide beta dose measurement into these categories: dose derived from spectral measurements, dose derived from “delayed” media such as thermoluminescent dosimeters and film, dose derived from “immediate” media such as scintillators and ion chambers, and dose derived from biological media such as hair diameter and skin erythema levels (ICRU 56 1994). This research involves dosimetry derived from scintillators of specific geometries.

Scintillation dosimeters may be categorized by function: those that are used in medical settings for measuring patient dose, those that are used in lab or work settings to measure occupational dose, and those developed for special research purposes.

In the medical setting, much of the effort has gone into dose measurements of high-energy photon beams [Beddar et al. (1992), de Boer et al. (1993), Mainardi et al. (1997), and Clift et al. (2000)]. Though not measuring beta dose, the materials are the same, namely plastic scintillators coupled to a light detector and associated electronics. The complications are also similar, for instance, the need to account for Cerenkov radiation. Not all efforts have been directed towards photon radiation therapy: Bambynek et al. (2000) developed a dosimetry system for cardiovascular brachytherapy beta sources using a plastic scintillator, Williamson et al. (1998) worked on plastic scintillator response to low-energy photons from brachytherapy sources, as did Kirov et al. (1999) and Flühs et al. (1996), and de Sousa et al. (2000) studied a dosimeter for patients undergoing diagnostic radiology procedures. The primary advantages of plastic

scintillator material in all of these cases are its near-water equivalence, a property useful when dose to tissue is desired, and small backscatter factors.

In the lab or workplace, a common technique used to measure beta dose is to first measure the beta spectrum with a scintillator, and then calculate a dose from that information. Martz et al. (1986) used a plastic scintillator 2.5 cm diameter by 0.9 cm deep to measure beta spectra and convert those spectra to dose. They used a beta energy deposition function, derived from calibrated sources, to convert the measured spectra to dose at a depth of 7 mg/cm². Thus, calculation of dose relied not only on direct extrapolation of scintillator light output to dose, but on previously derived calibration curves, in order to isolate the dose to a thin layer at a specific depth. Gammas were excluded by measuring spectra with and without a beta shield. Shen et al. (1987) used plastic scintillators to measure spectra, from which they subsequently calculated doses using electron transport theory as applied to TLDs. Swinth et al. (1989) constructed a combination proportional counter-plastic scintillation counter for measuring beta spectra and dose. They used coincidence gating to exclude gamma events. Dose was calculated from spectral information and compared to extrapolation chamber data for calibration. Horowitz et al. (1993) developed a two-detector telescope device consisting of a thin, front silicon detector and a thick, back plastic scintillator. Again, gamma rejection was accomplished by coincidence analysis. Dose was calculated by comparison to Monte Carlo depth distributions for the spectra measured. Vapirev et al. (1996) employed a plastic scintillator to measure beta spectra after passage of the betas through absorbers of various thicknesses. Dose was calculated via specific energy losses, dE/dx, taken from

ICRU 37, and the spectra. Results were compared to the calculations of Cross and Marr (1960).

Several authors have studied thin plastic scintillators for beta dosimetry. Bingo et al. (1980) developed a beta dose survey meter using a 2 mm thick scintillator. The premise was that there existed a certain thickness of scintillator that would satisfy a directly proportional relationship between count rate and dose rate, for all beta energies, i.e. independent of beta energy. Two millimeters happened to be the experimentally determined optimum thickness. Johnson et al. (1983) deliberately chose to use a very thin plastic scintillator, backed by a 1 cm thick Lucite light pipe, to measure dose to skin directly. Our device will be similar, except that we will study several configurations of scintillator layers of somewhat greater thickness, and most importantly we will use a large area avalanche photodiode instead of a photomultiplier tube.

Finally, on a somewhat esoteric note, Watt and Alkharam (1995) proposed using extremely thin (20 microns) plastic scintillators to directly simulate DNA damage, in the sense that the fluor spacing in the scintillator is analogous to the DNA diameter of around 2 nm. So, two scintillation emissions within 2 nm can be considered a double strand break, and thus an indication of dose.

For examples of techniques used to determine beta spectra independently of dose considerations, see Simons and Higginbotham (1990), who developed a beta

spectroscope with gamma discrimination capabilities and Palazzolo et al. (1992), who built a well-type plastic scintillator with gamma anticoincidence properties.

The theory of beta decay and the expressions predicting beta spectrum shape are covered in Evans (1955) and ICRU 56 (1994). Cross et al. (1983) have published beta energy-emission spectral shapes for about 100 nuclides.

For general consideration of scintillation theory and its application to radiation detection, see Knoll (1989). For insight into specific scintillator characteristics such as light yield, see Sysoeva et al. (2002), Moszynski et al. (1994, 1997) and Dorenbos et al. (1995). These papers provide valuable insight into experimental method.

Chapter 3: Materials and methods

Materials

Large area avalanche photodiode and electronics

The heart of the beta dosimetry and spectroscopy system is a module (Figure 1) built around a large area avalanche photodiode (LAAPD).



Figure 1: LAAPD module.

The LAAPD is analogous to a photomultiplier tube. Rather than the photocathode/dynode configuration common to PMTs, it's a photosensitive semiconductor with an avalanche region that converts the scintillation light to an amplified electrical signal. Unlike a PMT, the LAAPD module requires only a 12 Volt power supply, provided in this case by an Ortec NIM bin. The high voltage needed to create the avalanche region is provided by circuitry in the module. A potentiometer accessed by a small screw allows adjustment of the HV, and thus the gain. A digital panel meter, separate from the module and

constructed in-house, displays the HV. Output is provided through a BNC connection.

Operating specifications are detailed in Table 1.

Active diameter, mm	16
Spectral enhancement	Blue
Sensitivity @ 1 MHz (10^5 V/W)	3.9
NEP (pW/vHz)	0.43
Frequency range (MHz)	0.002 – 10
Feedback resistance (k Ω)	$10 \pm 1\%$
Output impedance (Ω)	40
Linear output swing (V)	+1
Output stability (%/°C)	± 1.2
Operating voltage (V)	± 12
Current at ± 12 V (mA)	20 - 150

Table 1: LAAPD properties.

The LAAPD is “blue enhanced”, meaning it has been designed to be sensitive towards lower wavelengths. Its peak sensitivity, however, is near 650 nm. This matches closely with the spectral output of our chosen scintillator.

The output signal is processed through a preamplifier (Ortec model 113), then through a shaping amplifier (Ortec model 460), and finally recorded in a multi-channel analyzer, Ortec Maestro. For certain studies the signal may also be fed through a pulse shaping

analyzer/timing SCA (Ortec model 552) and time-to-amplitude converter (Ortec model 567). These components, with the exception of the preamplifier, which is freestanding, are contained in the same NIM bin that provides power to the detector.

Scintillators

The scintillators are built of BC-430 plastic, provided by Saint -Gobain. BC-430 is advertised as a “red” scintillator, though the spectrum is peaked more in the orange-yellow region around 580 nm. It’s a double waveshifter scintillator: the original photons from the ionizing radiation interaction are absorbed by one shifter and emitted at a higher wavelength, which are in turn absorbed by a second shifter and emitted at the final, still higher, wavelength. This final spectrum matches the sensitive spectral region of the LAAPD reasonably well. Full specifications for BC-430 are summarized in Table 2.

Scintillator properties	
Light output, % Anthracene	45
Rise time, ns	3.2
Decay time, ns	16.8
Pulse width, FWHM, ns	17.7
Wavelength of maximum emission, nm	580
H atoms per cc	5.23×10^{22}
C atoms per cc	4.72×10^{22}
Ratio of H to C atoms	1.108
Electrons per cc	3.36×10^{23}

Scintillator properties	
Base	Polyvinyltoluene
Density	1.032 g/cc
Refractive index	1.58
Coefficient of linear expansion	7.8×10^{-5} , below 67 °C

Table 2: Properties of BC-430 plastic scintillator.

The scintillators were machined to our specifications, as follows. There are four separate scintillators, each cylindrical in shape with length 13 mm and diameter 15 mm. As the LAAPD diameter is 16 mm, this allows close coupling of the scintillator with the LAAPD. All surfaces save the output face, which is coupled to the LAAPD, are coated with reflective white paint.

Each cylinder has a distinct function. The first consists entirely of BC-430 and will be used to calibrate the light output, as well as for spectroscopic studies. The other three are for dosimetry studies and are of various configurations. One has a layer of BC-430, 1 mm thick, followed by 12 mm of transparent, inert BC-802 acrylic plastic. Another has 1 mm of BC-802 followed by 2 mm of BC-430 and 10 mm of BC-802. The last cylinder has 3 mm of BC-802 followed by 10 mm of BC-430. These last three cylinders are designed to demonstrate the capability of the device to measure beta dose at different levels of tissue. In principle, should this approach prove to be feasible, cylinders of nearly any configuration could be attached to the LAAPD, depending on need. Figure 2 illustrates the scintillator configurations currently under study.

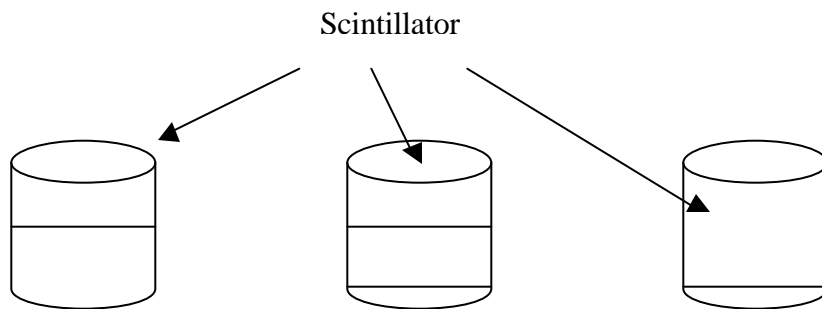


Figure 2: Scintillators.

Housing

The complete detector assembly will be housed in a construction that will enable us to position the sensitive detector face as close to the beta source as desired.

It consists of a tube that can be lowered or raised by means of a reversible ac electric motor controlled by a two-way switch. Cutoff switches at either end will prevent the tube rack from disengaging from the drive gear. The detector will be held motionless within the tube.

Sources

A final inventory of beta sources has yet to be assembled. However, it will be desirable to investigate as many as possible of the types commonly encountered. Point sources will be the primary focus of the research, though area sources will also be used, depending on availability. If the detector can accurately handle point sources, it should be able, by

extension, to handle area sources. In addition to ordinary beta emitters, a series of conversion electron sources will be used to calibrate the system. Candidates are ^{133}Ba , ^{207}Bi , and ^{137}Cs . Finally, a source such as ^{55}Fe , which emits a low-energy characteristic x-ray, will be needed to fully characterize the LAAPDs efficiency. Other nuclides, such as alpha emitters, may also be investigated.

Methods

Characterization of the LAAPD module

In order to fully understand the capabilities, strengths and limitations of the LAAPD as a radiation/scintillation detector, it is necessary to characterize it fully. This involves the following:

1. Establishing the sensitivity of the LAAPD by determining the number of electron-hole pairs produced in the LAAPD per MeV of energy deposited in the scintillator. This is done using methods described in the literature: by comparing the position of characteristic x-ray energy from ^{55}Fe deposited directly in the LAAPD with that from ^{137}Cs (or some other suitable gamma) in the scintillator, on the same scale, and knowing that 3.6 eV is required on average to produce one e-h pair, the number of e-h pairs produced in the LAAPD per MeV of energy deposited in the scintillator can be calculated. Knowing the number of photons produced per MeV in the scintillator allows us to calculate the quantum efficiency of the LAAPD *for the particular configuration at hand*.
2. Measuring the energy resolution of the system using conversion electrons incident on both the LAAPD and the scintillator.
3. Measuring the energy resolution of the system for soft x-rays.
4. Determination of the optimal gain so as to maximize the signal to noise ratio.
5. Determination of the change in gain as a function of temperature.

Characterization of the scintillator

The principal aspect of the scintillator is its efficiency in transforming ionizing radiation energy into optical radiation, or the number of photons produced per MeV. In this case the literature is relied upon. The scintillators are small enough to limit gamma interactions to a large extent, however, efforts will be made to screen out the gamma component if gamma contributions appear to be significant.

Calibration of the system

The energy scale of the device, as seen on the MCA, must be calibrated. This will be done using a range of conversion electrons, both via the scintillator and directly in the LAAPD.

Cerenkov component

For relativistic electrons interacting in certain materials, Cerenkov radiation can be a significant percentage of the light incident on a detector. Our device, with its water-equivalent components, falls into this category. For the dosimetry portion of this research, therefore, the Cerenkov component must be investigated. There are two approaches that will be studied: optical filtering and time filtering. A properly selected optical cutoff filter will eliminate the Cerenkov component while passing the non-Cerenkov portion virtually unchanged. Since Cerenkov radiation has a faster decay time than scintillation light, it can be removed by time filtering as well.

Theoretical dosimetry

In order to analyze the effectiveness of this device as a dosimeter, it is necessary to establish a baseline for comparison. To affect this comparison, several means of modeling beta dose will be utilized. These include empirical point source tables found in published works[Loevinger (1950, 1954, 1956), Berger (1974), ICRU (1994), Cross (1997)], standard textbook formulas such as are found in Martin (2000), computer models such as VARSKIN (Durham, 1993), and Monte Carlo calculations carried out for specific geometries [Shen et al. (1987), Hirayama (1994), Gualdrini and Padoani (1994)]. Point sources will be of primary interest, since other geometries can be derived from point source results.

Theoretical spectroscopy

As beta spectra have already been extensively calculated for all beta emitters of interest, published data will be relied on for comparison to experimental results.

Experimental dosimetry

Surface, shallow, and deep cases will be examined using the scintillators specifically designed for each. Point sources will be examined for a variety of beta emitters commonly encountered in radiation protection situations. Area sources will be examined if they become available. Results will be compared to the various theoretical calculations spelled out previously.

Experimental spectroscopy

Spectra for a variety of beta emitters will be measured, using the scintillator and the LAAPD directly. Results will be compared to the published data. Allowances will be made for the effects of air, scatter, and any other perturbing factors. In order to develop algorithms to identify mixed sources, combinations of beta emitters will also be measured. Development of these algorithms will likely fall along the lines of either a deconvolution process or a neural network solution. Most likely this will be someone else's effort, but it is expected that this device will provide raw data for those efforts.

Conclusion

A system for enhancing beta dosimetry and spectroscopy utilizing shaped plastic scintillators and a large area avalanche photodiode is proposed. Ultimately the device is envisioned as a field-ready beta detector, capable of identifying and quantifying beta sources present due to accidents or other actions.

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Appendix: Schedule

Task	2003				2004			
	<i>Spring</i>	<i>Summer</i>	<i>Fall</i>	<i>Winter</i>	<i>Spring</i>	<i>Summer</i>	<i>Fall</i>	<i>Winter</i>
Assemble Materials	X							
Characterize LAAPD: Sensitivity, quantum efficiency, energy resolution, gain, calibration.		X						
Characterize scintillator: Cerenkov, calibration.		X						
Theoretical dose calculations			X					
Dose measurements			X	X				
Beta spectrum measurements			X	X				
Data analysis				X	X			
Thesis writeup					X	X		
Defense							X	